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Single Degree of Freedom Representation of the Hybrid III Dummy and Cadaver Lower Limbs

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ABSTRACT

Dynamic axial impact tests conducted at the Medical College of Wisconsin on isolated cadaver and Hybrid III lower limbs were examined to characterize their dynamic response. Unembalmed cadaveric lower limbs, disarticulated at the knee, and the Hybrid III lower limb were fixed to a mini-sled system and axially impacted at the plantar surface of the foot by a pendulum at varying speeds and impact conditions. The axial impacts were along the distal one third of the tibia to minimize foot rotation. Load cells and uniaxial accelerometers were attached to the pendulum impactor and to the mini-sled. The lower limb attached to the mini-sled was modeled as a single degree of freedom system with a Kelvin element. The average stiffness and damping coefficients of the cadaveric and Hybrid III lower limbs were obtained using the measured accelerations on the pendulum and the mini-sled as input and output of the system, respectively. The average stiffness and damping coefficients were estimated to be 963 kN/m and 0.21 for the cadaver and 3256 kN/m and 0.26 for the Hybrid III dummy lower limbs, respectively.

INTRODUCTION

Recent research efforts have attributed lower limb injuries to axial loading through the plantar surface of the foot. In an epidemiological study using data from a Level I Trauma Center, Dischinger et al. (1994) and Crandall et al. (1996) noted that axial load through the plantar surface of the foot contributed to 47% of the ankle fractures sustained in frontal automobile crashes. In another study, Ziedler et al. (1981) noted that distal tibia and fibula fractures were caused by axial compression alone or by a combination of compression, torsion, bending, and tension. Injury criteria for the foot and ankle complex have recently been developed based on the contact axial force at the plantar surface of the foot (Klopp et al., 1997) and axial force in the proximal tibia (Yoganandan et al., 1996). These criteria were developed using human cadaver lower limbs.

In order to evaluate countermeasures, these injury criteria must be incorporated into testing with anthropomorphic test devices and computational models. However, before applying any lower limb injury criteria to the Hybrid III dummy, physical properties and response differences between the Hybrid III dummy and human cadaver lower limbs need to be addressed. In particular, the mass, stiffness, and damping properties of the dummy lower limb should be similar to that of the human such that the forces measured in the dummy leg and human leg are similar under similar impact conditions.

This paper presents the research efforts concerned with the application of a lower limb injury criteria based on axial force, to the Hybrid III dummy. The differences in the dynamic properties and the physical responses between the Hybrid III and human cadaver lower limbs were examined. The objective in this research effort was to determine the dynamic stiffness and damping properties of the human cadaver and Hybrid III dummy lower limb. In order to achieve this objective, the axial impact tests to the lower limbs of cadavers and Hybrid III dummy, conducted at the Medical College of Wisconsin (MCW) were examined (Yoganandan et al., 1996).

TEST SETUP AT THE MEDICAL COLLEGE OF WISCONSIN

The test apparatus (Figure 1) consisted of a pendulum and leg specimen, attached to a mini-sled. The mini-sled was free to move on linear ball bearings over precision ground stainless steel rails after the impact. The pendulum, mass of 25 kg, impacted the plantar surface of the foot with velocities ranging from 2.2 m/s to 5.6 m/s. The human cadaver leg specimens and the dummy leg were disarticulated at the knee and attached to the mini-sled. The mini-sled and leg assembly was ballasted to 16.8 kg in order to simulate the mass of the whole lower limb. Load cells and accelerometers were attached to the pendulum and the mini-sled system in order to measure accelerations and forces at the plantar surface of the foot and the proximal leg. The contact surface of the pendulum was padded with a one inch thick synthetic rubber (E.A.R. Composites Isodamp C-1000). Details of the test setup are provided in (Yoganandan et al., 1996).

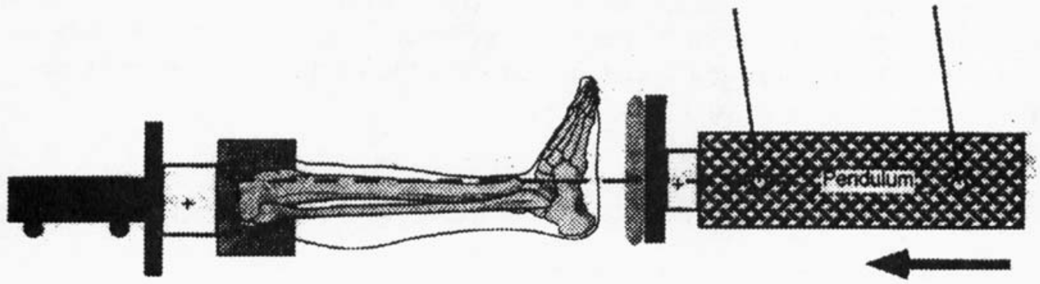


Figure 1. Setup for the axial impact tests at the Medical College of Wisconsin

PHYSICAL PROPERTIES AND RESPONSE DIFFERENCES BETWEEN THE HYBRID III DUMMY AND HUMAN CADAVER LOWER LIMBS

The average mass of the human lower limb was obtained from a summary of cadaver segment mass data (Crandall, 1996 and NASA, 1978). The average mass of the dummy lower limb was obtained from Dummy specifications. The mass of various segments of the lower extremity are presented in Table 1. The total mass of the Hybrid III dummy lower extremity (11.2 kg) is similar to the mass of the human cadaver lower extremity (11.5 kg).

Table 1. Mass of Segments of the Lower Extremity.

Segment	50% Hybrid III Dummy (kg)	Human Cadaver (kg)
Thigh	6.0	7.3
Leg	3.7	3.2
Foot	1.5	1.0
Total	11.2	11.5

In order to examine response differences, the Hybrid III dummy and cadaver lower limbs were impacted under identical conditions in the MCW test setup. Only those cadaver tests with no injury were considered. In each paired test, using Hybrid III and human cadaver lower limbs, the pendulum impact velocity and impact energy was maintained the same. For these tests, the dummy leg consistently experienced higher force (one and a half to two times higher) at the proximal end than the cadaver leg (Figure 2).

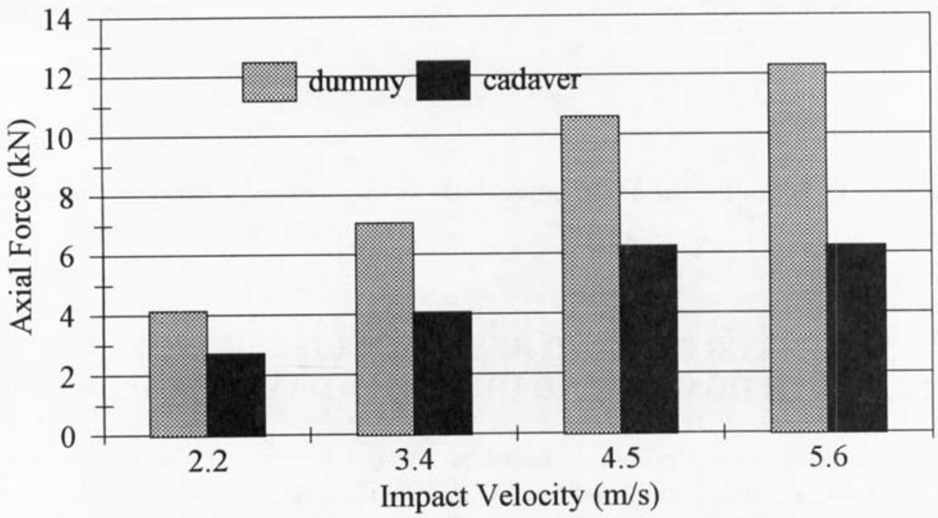


Figure 2: Cadaver and dummy responses under similar impact conditions

Since the mass of the Hybrid III dummy and cadaver lower limbs are similar, the results in Figure 2 imply that the dynamic stiffness and damping properties of the dummy and cadaver lower limbs differ. Due to the large response differences, the injury criteria developed using human cadaver lower limbs would be too conservative when applied to the Hybrid III lower limb.

Furthermore, a common countermeasure to reduce axial force in the leg is to apply padding to the floor pan structure. Due to the response differences between cadaver and dummy, the optimal padding stiffness selected would be different for the dummy and the cadaver lower limbs. For example, a padding which would reduce the force in the dummy leg by 40% may only reduce the force in the human leg by 20%.

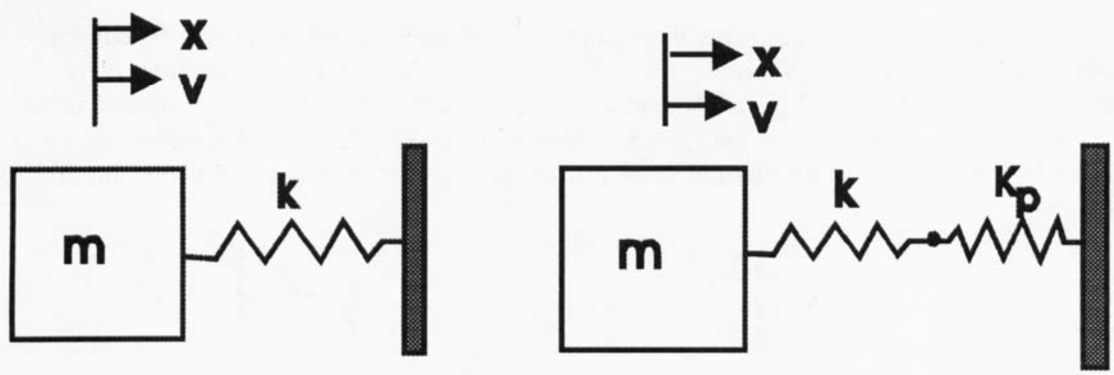
The ratio of axial force in the dummy leg to that in the cadaver leg under similar impact conditions (axial force response ratio) depends on the impact conditions. For illustration purposes, consider a simple spring-mass model to represent the lower limb (Figure 3a). The stiffness of the dummy and cadaver lower limbs are represented by k_d and k_c, respectively. The mass of the dummy and cadaver lower limbs are approximately the same, represented by m. For a purely linear, elastic impact with an initial velocity v, we have from conservation of energy principles:

$$\frac{1}{2}mv^2 = \frac{1}{2}k_d x_d{}^2 = \frac{1}{2}k_c x_c{}^2 \qquad (1)$$

$$k_d x_d = F_d \qquad and \qquad k_c x_c = F_c \qquad (2)$$

$$\Rightarrow \frac{F_d}{F_c} = \sqrt{\frac{k_d}{k_c}} \qquad (3)$$

where x_c and x_d are the relative displacements of the cadaver and dummy lower limbs.



(a) rigid impact surface **(b) padded impact surface**

Figure 3. Spring-mass models of lower limb and impact surface

The ratio of the force in the dummy leg with respect to that in the cadaver leg under the same impact condition, is the square root of the ratio of the dummy leg stiffness to the cadaver leg stiffness. Now, consider the situation where the impact surface is padded by a material of stiffness k_p Figure (3b). Using an equivalent stiffness for the dummy lower limb and padding (k_{de}), and for the cadaver lower limb and padding (k_{ce}), we have

$$k_{de} = \frac{k_p k_d}{k_p + k_d}, \qquad k_{ce} = \frac{k_p k_c}{k_p + k_c} \tag{4}$$

$$and \qquad \frac{F_d}{F_c} = \sqrt{\frac{k_{de}}{k_{ce}}} \tag{5}$$

The axial force response ratio between the dummy and cadaver leg different in the padded condition (equation 5) as compared to the unpadded condition (Equation 3). Therefore, the impact surface condition alters the scaling factor between the force in the dummy leg and cadaver leg.

In order to better understand the relationship between the axial force response ratio and the impact conditions, the first step was to characterize the dynamic properties of the dummy and cadaver lower limbs.

CHARACTERIZING THE DYNAMIC PROPERTIES OF THE CADAVER AND HYBRID III DUMMY LOWER LIMB

The stiffness and damping coefficients of the human cadaver and the Hybrid III dummy lower limbs were obtained by analyzing the MCW test data. The dummy and human cadaver lower limb were represented by a single degree of freedom system with a Kelvin element (Figure 4). The acceleration measured at the impactor surface of the pendulum was taken as input into the system. The acceleration measured at the proximal end of the leg was taken as the output of the system.

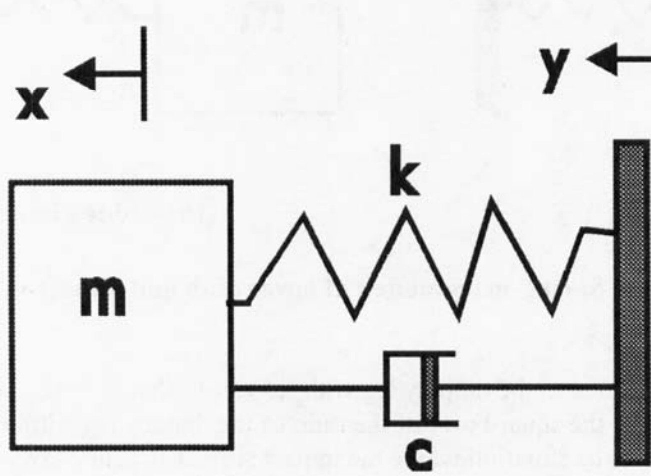


Figure 4. Single degree of freedom representation of the lower limb

The development of a single degree of freedom linear model required several simplifying assumption which are listed below:

1. The first assumption is that the stiffness and damping coefficient are linear. During the initial phase of impact, the foot penetrates the padding material producing a growing contact area. This produces a nonlinear force versus deformation response. In order to ensure a linear stiffness and damping coefficient, analysis of the data was considered only in the range between 5% of the peak value on the leading edge of the input pulse and 10% of the peak value on the trailing edge of the input pulse. Analysis of the MCW data suggested that within this range of data, linear approximation was reasonable.

2. The second assumption is that there is no significant foot rotation during the impact. Foot rotation would also produce a nonlinear response for axial measures. Hence, only those tests were considered from the Medical College of Wisconsin where the impact was along the distal one-third of the tibia to minimize foot rotation. Film analysis of these MCW tests showed little foot rotation.

3. The third assumption is that there is no loss of contact between the foot and the impacting surface during the impact event. This is a critical assumption of linearity since any loss of contact would be expected to produce a strongly nonlinear response. This means, that the data analysis is restricted to the duration of the compression pulse measured by the pendulum load cell. This assumption can also be satisfied by considering the signals only between the range of 5% of the peak value on the leading edge and 10% of the peak value on the trailing edge.

4. The fourth and last assumption is that the stiffness and damping properties of the system are constant during the compression phase. Any injury or fracture in a cadaver spesimen during the impact, would change its stiffness and damping properties. Hence, only those cadaver tests were considered in the analysis which did not sustain any injury.

For the single degree of freedom system in Figure 4, m represents the mass of the leg and mini-sled assembly (16.8 kg). The equivalent stiffness of the leg and padding on the pendulum surface is represented by k and is given by Equation 4 for the cadaver and Hybrid III dummy. Damping of the system is represented by c.

If $\ddot{y}$, $\dot{y}$, and y are the acceleration, velocity and displacement of the impacting surface, respectively, and $\ddot{x}$, $\dot{x}$, and x are the acceleration, velocity, and displacement at the proximal end of the leg, then the equation of motion for the system in Figure 4 is obtained from dynamic equilibrium for the lumped mass where

$$m\ddot{x} + c(\dot{x} - \dot{y}) + k(x - y) = 0 \tag{6}$$

$$Let \quad z = x - y$$

$$\rightarrow \quad m\ddot{z} + c\dot{z} + kz = -m\ddot{y} \tag{7}$$

$$or \quad \ddot{z} + 2\xi\omega\dot{z} + \omega^2 z = -\ddot{y} \tag{8}$$

$$rearranging, \quad -\ddot{x} = 2\xi\omega\dot{z} + \omega^2 z \tag{9}$$

The natural frequency of the system, ω, is √(k/m) and the damping coefficient, ξ, is c/2mω. The Equation 9 is a linear relation in z and $\dot{z}$ where these values are known for all corresponding $\ddot{x}$. This relation can be generalized for the measured data

$$-\ddot{x}_i = b_o + b_1\dot{z}_i + b_2 z_i + \varepsilon_i \tag{10}$$

$$where \quad b_1 = 2\xi\omega \quad and \quad b_2 = \omega^2$$

where $\ddot{x}_i, \dot{z}_i, z_i$ are the acceleration, relative velocity, and relative displacement of the system at the ith instant, b_o is a bias or offset value, and ε_i is a perturbation. The bias can be attributed to systematic errors in the measures. The perturbation is due in part to the precision of the instruments and to the fact that the true system response is not completely described by a single degree of freedom linear model. If one assumes that the perturbation is a random variable that is normally distributed, then an optimal line or line of best fit to the test data under consideration can be obtained by using multivariate linear regression. This method produces least squares estimates for the parameters b_0, b_1, b_2. The estimates of ω and ξ are then obtained from Equation 10. The stiffness and damping values are obtained from $k=\omega^2 m$ and $c=2\omega m\xi$. The mass considered is the mass of the leg and sled assembly (16.8 kg) which represents the total mass of the lower limb.

RESULTS

This method of linear regression was applied to each cadaver and dummy test. The errors in the estimates of b_1 and b_2 for each test were small (within 10%) and the R^2 value of the analysis was high (cadaver $R^2=0.9$ and dummy $R^2=0.98$) suggesting that the values of ω and ξ are reasonably constant during the compression phase of the impact. This implies that the assumptions of linearity and constant dynamic properties during the compression phase were reasonable. The details of the tests and the results of the regression analysis are presented in Table 2.

The average estimated stiffness and damping coefficient for the cadaver specimen is 963 kN/m and 0.21, respectively. The average stiffness and damping coefficient for the Hybrid III dummy is 3256 kN/m and 0.26, respectively. The results obtained are consistent with previous studies showing the dummy leg to be stiffer than the comparable size human leg (Crandall, et al., 1996). The axial stiffness of the cadaver lower limb is also consistent with Crandall's observation that the dynamic axial stiffness of the cadaver lower limb is approximately twice its static axial stiffness of 480 N/mm.

CONCLUSIONS

Axial impacts to the cadaver and Hybrid III lower limbs under similar impact conditions suggested that the axial force measured in the dummy leg is consistently higher than that measured in the cadaver leg. This difference in axial force response was attributed to the differences in dynamic stiffness and damping properties between cadaver and dummy lower limbs. The axial force response ratio between the dummy and cadaver lower limbs was shown to be dependent on the impact condition.

Axial impact tests conducted at the Medical College of Wisconsin were analyzed to estimate dynamic properties of the Hybrid III dummy and cadaver lower limbs. The lower limb was represented by a single degree of freedom linear system and a multivariate linear regression analysis was conducted on the MCW test data for this purpose. The average stiffness and damping coefficients were estimated to be 963 kN/m and 0.21 for the cadaver and 3256 kN/m and 0.26 for the dummy lower limbs, respectively. The dummy leg is more than three times stiffer than the cadaver leg and so under similar impact conditions experiences higher axial forces.

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Table 2: Test Data and Results of the Analysis

Test No.	Age	Sex	Test Velocity (m/s)	freq. ω Rad/s	damp. ratio ξ	stiffness kN/m	max. impactor force (N)	max. tibia force (N)
CADAVER TESTS								
PCLE110-1	27	M	2.2	235	0.19	928	3662	2666
PCLE114-1	46	M	2.2	242	0.15	984	3756	2719
PCLE116-1	27	M	3.4	282	0.22	1336	5919	4490
PCLE117-1	55	M	4.5	277	0.21	1289	7819	6226
PCLE129	66	M	3.4	230	0.22	889	5900	4607
PCLE131	64	M	2.2	206	0.18	713	3564	2776
PCLE132	50	M	3.4	209	0.3	734	4236	3093
PCLE133	50	M	5.6	223	0.24	835	7983	6239
AVERAGE				238	0.21	963		
DUMMY TESTS								
PDLE119-1			2.2	356	0.2	2129	5447	4125
PDLE119-2			3.4	440	0.25	3252	9434	7058
PDLE119-3			4.5	498	0.27	4166	13960	10604
PDLE119-4			5.6	455	0.32	3478	17840	12278
AVERAGE				437	0.26	3256		

DISCUSSION

PAPER: **Single Degree of Freedom Representation of the Hybrid III Dummy and Cadaver Lower Limbs**

PRESENTER: Shashi Kuppa, Conrad Technologies, Inc.

QUESTION: Barry Myers, Duke University

Nice piece of work and good luck with what never works when you really look at it. I'm a little concerned about your choice of a model. You've got a Kelvin solid in front of a mass and the more your input looks like a step function, the more that system blows up. In other words, you've got a damper locked between your input and the mass. A damper moving instantaneously generates infinite force, and so I think one of the problems is that the higher the frequency content of your input, the more that system is going to lock up and I don't think the body does that. So, you may find that you have a model that is a little more complex but actually much better behaved over frequency if you use a three parameter solid instead of a Kelvin solid in front of your mass. Because that damper is going to lock up, what will happen when you regress it is that you will end up with a damping ratio that is too low because the damper will harden up your system at faster frequencies. I think the data that you validated against was relatively unpadded, so I think your damper may be getting pushed down in your regression model.

A: It wasn't only that. We did see nonlinearities in the beginning, so we were starting our linear regression process at around ten percent of the forces, not from the beginning.

Q: That is probably all the soft tissue compressing before you get into the bone.

A: That is why, because it is the heal pad that compresses first.

Q: It is probably worth the time to just run a different viscoelastic model in front of the mass and see if it doesn't help you at all.

A: OK. Thank you.

Q: Guy Nusholtz, Chrysler Corporation

Padding is inherently non-linear. In fact, for all practical purposes, it goes up and then flat tops, depending on how you choose your padding. Some have second slopes, others are flat. Some will go up and drop down and so when you are trying to do linear regression where you are just using a linear spring ...

A: I had the padding characteristics from the manufacturer, which was very linear for that range.

Q: What type of padding are you using; was it just a rubber?

A: It is a synthetic rubber padding made by E.A.R.

Q: Most padding in small compression is linear; it is just a straight line. However, it becomes nonlinear as soon as you get any type of compression, which is what happens in a floor pan.

A: We are still working on this process now, and I don't know how to address the nonlinearities in a floor pan. For the linear regression process, I think our padding behaved fairly linear for that condition. It was a very rigid padding.

Q: So, in that small range that you have calibrated it, you are showing under those conditions, that you get an increased stiffness. But, when you start going into a lot of different environments, it may not extrapolate the same way as just a linear system and you may need to look at other types of approaches to characterize the stiffness. When you start to move out of that very narrow range where you calibrated that system, you may discover that your approach doesn't work. Although, you can probably believe that the dummy leg is always going to be stiffer to some degree.

A: Thank you.

Q: Craig Morgan, Denton

The latest version of the Hybrid III foot includes a foam insert in the heel. Did you use that?

A: No, we didn't use that one.

Q: OK. Thank you.